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In Vitro Temperature Change at the Dentin/Pulpal Interface by Using Conventional Visible Light Versus Argon Laser

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Background and Objective: **The argon laser has been promoted as a competing technology to multi-wavelength visible light as a curing source for dental restorative resins. However, the comparative thermal risk to the pulp between these two sources of light energy requires determination. The objective of this study is to compare the temperature induced at the dentin-pulpal interface between the argon laser and visible light curing unit at a variety of exposure regimens and conditions.**
Study Design/Materials and Methods: **In vitro temperatures were measured and recorded at the dentin-pulpal interface upon external light exposure. Independent variables included the dentin thickness, duration and waveform of exposure, and presence of composite resin.**
Results: **In most instances, the argon laser resulted in less temperature rise on the pulpal-dentin interface.**
Conclusion: **The argon laser should not pose a serious thermal risk to the pulp if used at recommended energies. Lasers Surg. Med. 26:386–397, 2000**

Key words: laser; polymerization; heat transfer; dental pulp

INTRODUCTION

Adequate polymerization is a crucial factor in obtaining optimal clinical performance of dental composite resin restorations. Problems associated with inadequate polymerization include inferior physical properties, decreased retention, and increased solubility in the oral environment. In addition, the presence of a substantial quantity of unpolymerized monomers in the resin may produce an adverse pulpal response [1].

Current visible light curing units often fail to meet the challenges of today's demands for more complex resin restorations or for the curing of resin cements through indirect esthetic restorations. Inadequate power output, long curing times, narrow light tips (<12 mm diameter), and the degradation of components (bulbs, reflectors, filters, and light tips) that occurs with long-term use of the unit make it difficult to adequately cure composite resin, especially in deeper areas [2]. The degree of effective polymerization of the internal portion of a light-activated composite resin decreases as the distance from the light source increases and the presence of a cured surface does not ensure adequate polymerization throughout the restoration [3].

One concern with the use of the argon laser to photoinitiate resin polymerization is the potential for damage to the dental pulp and periodontal soft tissue resulting from heat produced during the curing process. Previous studies investigated the physical properties and polymerization stresses associated with the curing of composite

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restorative material by using an argon laser [4–8]. A few reports have looked at thermal effects on the pulp and changes in tooth structure as a result of argon laser irradiation of the surface of teeth. However, there is a lack of information regarding thermal changes at the pulpal wall when the surface of a tooth is exposed to the argon laser through defined thicknesses of dentin, cavity preparations, and when curing composite placed in the preparations. The objective of this study was to compare the differences in temperature at the pulpal dentin interface resulting from the use of an argon laser versus a conventional visible light curing unit to expose (1) a dentin surface, (2) a dentin surface containing a standard cavity preparation, and (3) to photopolymerize a resin restoration within the cavity preparation.

MATERIALS AND METHODS

An Optilux 401 visible light curing (VLC) unit (Demetron Research Corp., Danbury, CT) and an argon laser, model 5500 A (ILT Systems, Salt Lake City, UT) were used as the comparative light sources for this study. The VLC unit is a gun-type unit with a halogen lamp bulb, with a power output in the range of 450–500 mW. This VLC unit has emission lines that range from 400–520 nm with a bandwidth of approximately 120 nm. The wand tip diameter was 13 mm. The actual power density based on a 500 mW output was calculated to be 376 mW/cm^2. This rate resulted in an energy density of 60 J/cm^2 for the total exposure time of 160 seconds. To ensure uniform intensity, the curing light performance was monitored by use of an Ophir laser power meter (Ophir Optronics, Inc, North Reading, MA) before each research session.

The argon laser used in this experiment is a prototype unit designed for investigative use that permits variation of power. It has emission lines with wavelengths of 454.6 nm, 457.9 nm, 465.8 nm, 472.7nm, 476.5 nm, 488 nm, 496.5 nm, and 501.7 nm and a bandwidth of 47.1 nm. The system has a maximal rated continuous output power of 300 mW. The actual power output of the argon laser in this study was confirmed to be 285 mW by an Ophir laser power meter at the beginning of each research session. The spot size of the light beam emitted from the laser was 6 mm in diameter, resulting in a power density of 1,000 mW/cm^2. The energy density for the total exposure time of 20 seconds was 20 Joules/cm^2.

The composite resin used in this study was

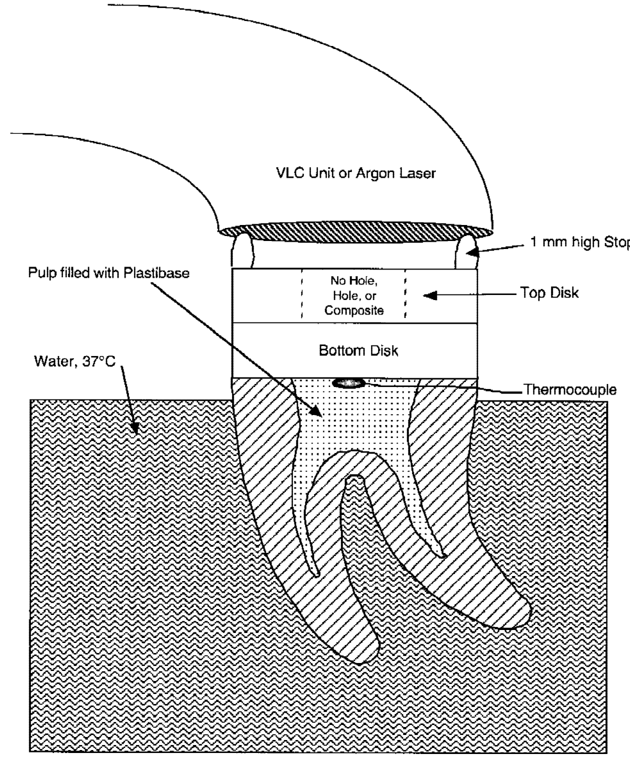

Fig. 1. Diagram of experimental set-up. VLC, visible light curing.

Durafill VS, shade C-10, Batch 027 (Kulzer, Germany). This resin is a microfill composite with silica filler particles of <0.04 µm, and a Bis-GMA resin matrix. The volume percent of filler particles is less than 50%. Camphoroquinone is the main photoinitiator.

A diagram of the general physical configuration tested is found in Figure 1. The base of the experimental tooth specimen was an intact extracted human third molar with noncarious and cleaned surfaces. The top part of the crown was sectioned and removed, exposing the pulp chamber. Dentin disk specimens of 1-mm and 2-mm thickness were also cut and polished to fit with 600-grit SiC paper. All specimens were stored in 0.2% solution of thymol in distilled water until testing. The pulp chamber of the base tooth was cleaned and filled with Plastibase (Squibb), a silicate compound with thermal properties similar to pulp tissue [9]. During testing, the disks were placed on the base tooth and sealed with Plastibase to prevent air from getting between the dentin disk and base tooth, which would create thermal discontinuity.

Temperature change was detected at the pulpal-dentin interface by using a thermocouple (Copper-Constantan 0.005-inch bead diameter,

Omega Engineering, Inc., Stamford, CT). The distal ends of the VLC and laser tips were consistently placed 1 mm from the specimen surface by resting on two composite stops, 1-mm thick, bonded to the periphery of the dentin disk at opposite ends. The root surfaces and the lower portion of the crown of the base tooth were partially submerged in a water bath during the testing procedure. This method effectively stabilized the internal baseline temperature at 37°C ± 0.5°C and was done to minimize the effects of ambient temperature changes and to provide a consistent initial temperature for each data set (Fig. 1).

Temperature change was recorded at baseline and at every second during the curing cycles by using the thermorecording software program DATALOG.exe version 4.30 developed at the University of Iowa College of Medicine. The temperature changes were also measured during a 1-minute cool down period after the exposures.

The duration of one curing cycle for the argon laser was 5 seconds, per manufacturer's recommendations at the time of this study. With respect to completeness of cure, this timing had been reported to be comparable to a 40-second curing cycle, which was recommended for the VLC unit [10]. Each specimen was exposed to four curing cycles by each of the light sources to simulate a worst-case clinical situation, wherein a given restoration is exposed to four curing cycles. The exposures were carried out in two sequences. First, the light was applied continuously for four times the manufacturer's recommendation to determine maximal thermal change. Second, to more closely simulate a clinical experience, the light was applied in an interrupted mode with a period of 20 seconds off-time between each of four exposures. Respectively, the exposure times were 5 and 20 seconds for the argon laser and VLC source. In those cases for which composite was placed and a continuous exposure used, measurement began immediately upon placement of composite. For the interrupted exposure sequence, the composite was placed after the second exposure to simulate curing the adhesive and then the composite in the clinical situation.

The sequencing of the test configurations and their abbreviations are shown in Table 1. Six data runs were performed for each of the 36 configurations. Initially, a 1-mm intact dentin disk was placed over the Plastibase-filled pulp chamber containing the thermocouple and a series of six data runs were recorded for each light at maximal exposure times for both continuous and interrupted cycles. Another sequence of data runs were recorded by using the same experimental design but with the addition of a 1-mm dentin disk placed on top of the base disk. A third set of runs simulated a cavity preparation by cutting a hole 3 mm in diameter in the center of the upper disk. And lastly, a set of runs was completed with composite placed in this hole and polymerized during the light exposure process, to observe the contribution from the polymerization exotherm to the temperature increase. These runs were then repeated by using a 2-mm-thick upper dentin disk. And finally, the entire process was repeated by using a 2-mm base dentin disk. This resulted in a total of 216 data sets (see Table 1).

A one-way analysis of variance (ANOVA) and Duncan's Multiple Range tests for pairwise contrasts were used for analysis of the data. A confidence level of 99% ($\alpha = 0.01$) was predetermined to be statistically significant.

RESULTS

The mean, standard deviation, and differences in temperature rise at the pulpal dentin interface for the argon laser and the VLC unit for various conditions are listed in Tables 2 through 10. Differences in both maximal temperature rise and temperature rise at specified exposure times were compared. ANOVA demonstrated statistically significant differences ($P = 0.01$) between mean temperature rise for the conventional visible light curing source and the argon laser and for continuous versus interrupted cycles for the same light source for comparable pairs of test tooth configurations.

Table 2 compares the mean maximal temperature rise for the argon laser and the VLC unit for the nine test tooth configurations, by using continuous exposure at four times the manufacturer's recommendation. In all configurations, temperature increase was significantly higher for the VLC unit than for the argon laser ($P = 0.01$). The highest temperature rise, 15.00°C, resulted from exposure of the VLC unit through a 1-mm base dentin disk (O1M). This compared with a 9.05°C rise with the argon laser (A1M) for a difference of 5.95°C.

Table 3 provides results of the mean maximal temperature rise for the same nine pairs of comparable tooth configurations, by using an interrupted exposure sequence for both the argon laser and the VLC unit, at four times the manufacturer's recommendation. Again, temperature

TABLE 1. Test Configurations*

Run sequence	Code	No. of runs	Disk thickness (mm) Bottom disk	Disk thickness (mm) Top disk	Hole status of top disk None	Hole status of top disk Empty	Hole status of top disk Filled	Exposure Cont.	Exposure Int.	Light source VLC	Light source Argon laser
1	O1M	6	1	None				x		x	
2	A1M	6	1	None				x			x
3	PO1M	6	1	None					x	x	
4	PA1M	6	1	None					x		x
5	O1M1H	6	1	1		x		x		x	
6	A1M1H	6	1	1		x		x			x
7	PO1M1H	6	1	1		x			x	x	
8	PA1M1H	6	1	1		x			x		x
9	O1M1C	6	1	1			x	x		x	
10	A1M1C	6	1	1			x	x			x
11	PO1M1C	6	1	1			x		x	x	
12	PA1M1C	6	1	1			x		x		x
13	O1M2C	6	1	2			x	x		x	
14	A1M2C	6	1	2			x	x			x
15	PO1M2C	6	1	2			x		x	x	
16	PA1M2C	6	1	2			x		x		x
17	O2M	6	2	None				x		x	
18	A2M	6	2	None				x			x
19	PO2M	6	2	None					x	x	
20	PA2M	6	2	None					x		x
21	O2M1H	6	2	1		x		x		x	
22	A2M1H	6	2	1		x		x			x
23	PO2M1H	6	2	1		x			x	x	
24	PA2M1H	6	2	1		x			x		x
25	O2M1C	6	2	1			x	x		x	
26	A2M1C	6	2	1			x	x			x
27	PO2M1C	6	2	1			x		x	x	
28	PA2M1C	6	2	1			x		x		x
29	O2M2H	6	2	2		x		x		x	
30	A2M2H	6	2	2		x		x			x
31	PO2M2H	6	2	2		x			x	x	
32	PA2M2H	6	2	2		x			x		x
33	O2M2C	6	2	2			x	x		x	
34	A2M2C	6	2	2			x	x			x
35	PO2M2C	6	2	2			x		x	x	
36	PA2M2C	6	2	2			x		x		x

*Total: 216.

Code: O, Optilux 401 visible light from Demetron; A, argon laser; P, interrupted, 20 seconds off-time between exposures; 1M, 1-mm-thick base dentin disk; 2M, 2-mm-thick base dentin disk; 1H, 1-mm top dentin disk with 3-mm diameter hole; 2H, 2-mm-thick base dentin disk with 3-mm diameter hole; 1C, 1-mm-thick base dentin disk with composite in hole; 2C, 2-mm-thick base dentin disk with composite in hole. Cont., continuous; Int., intermittent; VLC, visible light cured.

rise was consistently higher for the VLC unit, with 13.08°C through a 1-mm dentin disk being the highest (PO1M). A 4.88°C rise was found for the same configuration by using the argon laser (PA1M). Statistical analysis determined that significant differences ($P = 0.01$) in maximal temperature rise existed for all nine test tooth pairs. Again, temperature increase by using the argon laser was always less than for the VLC unit.

Tables 4 and 5 compare the temperature rise at the pulpal dentin interface by using the same light source in a continuous or interrupted mode. Table 4 contains results comparing the mean maximal temperature rise of the argon laser, by using either the continuous or the interrupted exposure sequence, at four times the manufacturer's recommendation. Statistical analysis between comparable pairs indicated a significantly higher ($P = 0.01$) mean maximal temperature rise for the continuous vs. interrupted argon light in six out of the nine test tooth configurations. Although the temperature increase was slightly higher for the interrupted vs. the continuous argon laser for A1M1C vs. PA1M1C and A2M2C vs. PA2M2C, the interrupted argon laser exposure produced a significantly higher temperature rise than did the continuous exposure only for A1M2C vs. PA1M2C. The highest temperature rise was

TABLE 2. Temperature Rise (°C) at Pulpal Dentin Interface, Argon Laser vs. Optilux VLC Unit* (Continuous Exposure)

Sample	Max temp rise	Difference	% Difference	*F* value	Pr > *F*
A1M	9.05 ± 0.91	5.95	40	173.40	0.0001
O1M	15.00 ± 0.63				
A1M1H	5.63 ± 0.19	1.80	24	187.46	0.0001
O1M1H	7.43 ± 0.26				
A1M1C	4.37 ± 0.49	3.23	42	189.63	0.0001
O1M1C	7.60 ± 0.30				
A1M2C	3.12 ± 0.27	5.10	62	483.31	0.0001
O1M2C	8.22 ± 0.50				
A2M	5.03 ± 0.30	3.05	38	33.78	0.0002
O2M	8.08 ± 1.25				
A2M1H	4.10 ± 0.33	5.23	56	902.74	0.0001
O2M1H	9.33 ± 0.27				
A2M1C	2.93 ± 0.44	5.30	64	246.55	0.0001
O2M1C	8.23 ± 0.70				
A2M2H	3.30 ± 0.50	4.05	55	259.26	0.0001
O2M2H	7.35 ± 0.36				
A2M2C	1.90 ± 0.52	6.18	76	377.64	0.0001
O2M2C	8.08 ± 0.58				

*Note: Duration of light exposure is four times the manufacturer's recommendation. Argon laser: 5 seconds × 4; total light exposure of 20 seconds. Optilux 401 VLC unit: 40 seconds × 4; total light exposure of 160 seconds.

9.05°C for the continuous argon laser, again through the 1-mm dentin disk (A1M). The highest temperature rise for the interrupted argon laser was 4.88°C, found for the PA1M configuration through 1-mm base dentin disk.

The data in Table 5 are the mean maximal temperature rise for the VLC unit by using either the continuous or the interrupted sequence, at four times the manufacturer's recommended exposure time. Statistical analysis between similar pairs indicated a significant difference (P = 0.01) in mean maximal temperature rise for seven of the nine tooth configurations, with temperature increase being greater for the continu-

TABLE 3. Temperature Rise (°C) at Pulpal Dentin Interface, Argon Laser vs. Optilux VLC Unit* (Interrupted)

Sample	Max temp rise	Difference	% Difference	*F* value	Pr > *F*
PA1M	4.88 ± 0.85	8.2	63	332.73	0.0001
PO1M	13.08 ± 0.70				
PA1M1H	3.60 ± 0.39	2.53	41	166.62	0.0001
PO1M1H	6.13 ± 0.28				
PA1M1C	4.83 ± 0.58	3.90	50	235.28	0.0001
PO1M1C	9.04 ± 0.34				
PA1M2C	3.88 ± 0.50	3.08	49	112.32	0.0001
PO1M2C	7.78 ± 0.75				
PA2M	3.15 ± 0.35	3.85	57	115.08	0.0001
PO2M	6.23 ± 0.61				
PA2M1H	2.85 ± 0.36	3.66	57	181.65	0.0001
PO2M1H	6.70 ± 0.60				
PA2M1C	2.12 ± 0.34	3.66	63	232.29	0.0001
PO2M1C	5.78 ± 0.48				
PA2M2H	2.05 ± 0.37	4.62	69	284.28	0.0001
PO2M2H	6.67 ± 0.56				
PA2M2C	2.60 ± 0.48	3.73	59	265.43	0.0001
PO2M2C	6.33 ± 0.29				

*Note: Duration of light exposure is four times the manufacturer's recommendation with 20-second intervals between consecutive exposures. Argon laser: 5 seconds on, 20 seconds off × 3, then 5 seconds on; total light exposure of 20 seconds. Optilux 401 VLC unit: 20 seconds on, 20 seconds off × 2, then 40 seconds on, 20 seconds off × 2, then 40 seconds on; total light exposure of 160 seconds.

TABLE 4. Temperature Rise (°C) at Pulpal Dentin Interface, Argon Laser* (Continuous vs. Interrupted)

Sample	Max temp rise	Difference	% Difference	*F* value	Pr > *F*
A1M	9.05 ± 0.91	4.17	46	67.29	0.0001
PA1M	4.88 ± 0.85				
A1M1H	5.63 ± 0.19	2.03	36	131.38	0.0001
PA1M1H	3.60 ± 0.39				
A1M1C	4.37 ± 0.49	−0.46	−11	2.20	0.1686
PA1M1C	4.83 ± 0.58				
A1M2C	3.12 ± 0.27	−0.76	−24	10.73	0.0083
PA1M2C	3.88 ± 0.50				
A2M	5.03 ± 0.30	1.88	37	99.79	0.0001
PA2M	3.15 ± 0.35				
A2M1H	4.10 ± 0.33	1.25	30	39.31	0.0001
PA2M1H	2.85 ± 0.36				
A2M1C	2.93 ± 0.44	0.81	28	12.73	0.0051
PA2M1C	2.12 ± 0.34				
A2M2H	3.30 ± 0.50	1.25	38	24.23	0.0006
PA2M2H	2.05 ± 0.37				
A2M2C	1.90 ± 0.52	−0.70	−36	5.87	0.0359
PA2M2C	2.60 ± 0.48				

*Note: Duration of light exposure is four times the manufacturer's recommendation. Argon laser (continuous): 5 seconds × 4; total light exposure of 20 seconds. Argon laser (interrupted): 5 seconds on, 20 seconds off × 3, then 5 seconds on; total light exposure of 20 seconds.

ous vs. interrupted exposure cycle in all but one of these nine pairs. As in all other comparisons, the greatest temperature increase was found with exposure through 1 mm of dentin. This was 15.00°C for the continuous (O1M) compared with 13.08°C for the interrupted VLC unit cycle (PO1M) for a difference of 1.92°C. No significant difference ($P > 0.01$) existed for the comparable pairs O1M2C vs. PO1M2C and O2M2H vs. PO2M2H, although again the temperature increase was greater for the continuous vs. interrupted exposure cycle.

Tables 6–8 provide the results of the analysis of the mean temperature rise at the pulpal dentin

TABLE 5. Temperature Rise (°C) at Pulpal Dentin Interface, Optilux VLC Unit* (Continuous vs. Interrupted)

Sample	Max temp rise	Difference	% Difference	*F* value	Pr > *F*
O1M	15.00 ± 0.63	1.92	13	24.94	0.0005
PO1M	13.08 ± 0.70				
O1M1H	7.43 ± 0.26	1.30	17	69.45	0.0001
PO1M1H	6.13 ± 0.28				
O1M1C	7.60 ± 0.30	−1.44	−19	60.51	0.0001
PO1M1C	9.04 ± 0.34				
O1M2C	8.22 ± 0.50	0.44	5	1.43	0.2594
PO1M2C	7.78 ± 0.75				
O2M	8.08 ± 1.25	1.85	23	10.61	0.0086
PO2M	6.23 ± 0.61				
O2M1H	9.33 ± 0.27	2.63	28	95.87	0.0001
PO2M1H	6.70 ± 0.60				
O2M1C	8.23 ± 0.70	2.45	30	49.99	0.0001
PO2M1C	5.78 ± 0.48				
O2M2H	7.35 ± 0.36	0.68	9	6.26	0.0313
PO2M2H	6.67 ± 0.56				
O2M2C	8.08 ± 0.58	1.75	22	43.70	0.0001
PO2M2C	6.33 ± 0.29				

*Note: Duration of light exposure is four times the manufacturer's recommendation. Optilux 401 VLC unit (continuous): 40 seconds × 4; total light exposure of 160 seconds. Optilux 401 VLC unit (interrupted): 20 seconds on, 20 seconds off × 2, then 40 seconds on and 20 seconds off × 2, then 40 seconds on; total light exposure was 160 seconds.

TABLE 6. Temperature Rise (°C) at Pulpal Dentin Interface, Argon Laser vs. Optilux VLC Unit* (Continuous) (Argon Laser –5 sec vs. Optilux –40 sec)

Sample	Temp rise (5 vs. 40)	Difference	% Difference	F value	Pr > F
A1M	5.92 ± 0.39	4.00	40	263.95	0.0001
O1M	9.92 ± 0.46				
A1M1H	3.07 ± 0.31	1.58	34	94.44	0.0001
O1M1H	4.65 ± 0.25				
A1M1C	1.68 ± 0.57	3.04	64	160.96	0.0001
O1M1C	4.72 ± 0.14				
A1M2C	1.10 ± 0.29	3.27	75	280.78	0.0001
O1M2C	4.37 ± 0.38				
A2M	2.70 ± 0.27	2.03	43	28.58	0.0003
O2M	4.73 ± 0.89				
A2M1H	2.05 ± 0.36	2.83	54	159.04	0.0001
O2M1H	4.43 ± 0.29				
A2M1C	1.17 ± 0.27	3.36	75	271.71	0.0001
O2M1C	4.53 ± 0.42				
A2M2H	1.72 ± 0.28	1.78	51	182.79	0.0001
O2M2H	3.50 ± 0.16				
A2M2C	0.57 ± 0.30	3.11	84	368.23	0.0001
O2M2C	3.68 ± 0.26				

*Note: Duration of light exposure is one time the manufacturer's recommendation. Argon laser: 5 seconds. Optilux 401 VLC unit: 40 seconds.

interface for the argon laser at one, two, and four times the manufacturer's recommended exposure time (5, 10, and 20 seconds, respectively). These results were then compared with the manufacturer's recommended exposure time (40 seconds) for the VLC unit by using continuous exposure.

Table 6 compares the mean temperature rise of the argon laser at 5 seconds vs. the VLC unit at 40 seconds. Statistical analysis between pairs indicated a significantly higher (P = 0.01) mean temperature rise for the VLC unit vs. argon light for all of the nine test tooth configurations. The highest temperature rise was again found with the O1M vs. A1M configuration: 9.92°C for the VLC unit and 5.92°C for the argon laser.

Results in Table 7 compare the mean temperature rise for the argon laser vs. VLC by using continuous exposure at 10 and 40 seconds, respectively. Temperature rise was significantly higher for the VLC unit vs. argon light source (P = 0.01) in all but the A2M vs. O2M configuration. The highest temperature rise was again found with the O1M vs. A1M configuration at 9.92°C for the VLC unit and 7.45°C for the argon laser.

Table 8 compares mean temperature rise for the argon laser at 20 seconds vs. the VLC unit at 40 seconds by using continuous exposure. Although the VLC unit had higher temperature rise in seven of nine cases, results were inconsistent, with four of the nine pairs demonstrating a significant difference. In one of these pairs, the argon laser had a greater temperature rise and in the other three pairs, the temperature increase with the VLC unit system was higher. The highest temperature rise was found for the 1-mm base dentin disk, with the VLC unit at 9.92°C compared with 9.05°C for the argon laser.

In Table 9, the mean temperature rise at the pulpal dentin interface was compared for the argon laser at 20 seconds and the VLC unit at 80 seconds. Consistently higher temperature rise was found for the VLC unit vs. the argon laser. All differences were statistically significant (P = 0.01). The highest temperature rise was again found by using the VLC unit through a 1-mm base dentin disk that was 13.08°C compared with 9.05°C for the argon laser.

Results in Table 10 demonstrate the insulating effect of increasing dentin thickness between the pulp and exposed surface. Temperature rise with the argon laser was compared for 1-mm and 2-mm base dentin at maximal exposure times for both continuous and interrupted cycles. The same protocol was followed with the VLC unit. For all samples, the temperature elevation through 2 mm of dentin was less than that for the 1-mm dentin thickness and all differences were significant. For most pairs, temperature rise through 1 mm of dentin was nearly double that found through 2 mm of dentin, which suggests an in-

TABLE 7. Temperature Rise (°C) at Pulpal Dentin Interface, Argon Laser vs. Optilux VLC Unit* (Continuous) (Argon Laser –10 sec vs. Optilux –40 sec)

Sample	Temp rise (10 vs. 40)	Difference	% Difference	F value	Pr > F
A1M	7.45 ± 0.68	2.47	25	54.31	0.0001
O1M	9.92 ± 0.46				
A1M1H	3.97 ± 0.40	0.68	15	12.47	0.0054
O1M1H	4.65 ± 0.25				
A1M1C	2.77 ± 0.52	1.95	41	68.67	0.0001
O1M1C	4.72 ± 0.14				
A1M2C	1.98 ± 0.29	2.49	57	149.99	0.0001
O1M2C	4.37 ± 0.38				
A2M	3.73 ± 0.26	1.00	21	6.98	0.0247
O2M	4.73 ± 0.89				
A2M1H	2.88 ± 0.44	1.55	35	51.91	0.0001
O2M1H	4.43 ± 0.29				
A2M1C	1.80 ± 0.45	2.73	60	118.2	0.0001
O2M1C	4.53 ± 0.42				
A2M2H	2.38 ± 0.39	1.12	32	42.35	0.0001
O2M2H	3.50 ± 0.16				
A2M2C	1.03 ± 0.27	2.65	72	299.89	0.0001
O2M2C	3.68 ± 0.26				

*Note: Duration of light exposure is two times the manufacturer's recommendation for the Argon laser, i.e., 10 seconds. Optilux 401 VLC unit is one time the manufacturer's recommendation, i.e., 40 seconds.

verse relationship between dentin thickness and temperature increase.

A typical example of the time vs. temperature profiles are shown in Figures 2 and 3. Figure 2 depicts O1M vs. PO1M and Figure 3 A1M vs. PA1M. Temperature readings were plotted every second from 20 seconds baseline and through the light application cycle, which started at 30 seconds and for 1 minute after the light was removed from the sample.

TABLE 8. Temperature Rise (°C) at Pulpal Dentin Interface, Argon Laser vs. Optilux VLC Unit* (Continuous) (Argon Laser –20 sec vs. Optilux –40 sec)

Sample	Temp rise (20 vs. 40)	Difference	% Difference	F value	Pr > F
A1M	9.05 ± 0.91	0.87	9	4.37	0.0631
O1M	9.92 ± 0.46				
A1M1H	5.63 ± 0.19	–0.98	–21	58.44	0.0001
O1M1H	4.65 ± 0.25				
A1M1C	4.37 ± 0.49	0.35	7	2.83	0.1234
O1M1C	4.72 ± 0.14				
A1M2C	3.12 ± 0.27	1.25	29	43.14	0.0001
O1M2C	4.37 ± 0.38				
A2M	5.03 ± 0.30	–0.30	–6	0.61	0.4521
O2M	4.73 ± 0.89				
A2M1H	4.10 ± 0.33	0.33	7	3.39	0.0956
O2M1H	4.43 ± 0.29				
A2M1C	2.93 ± 0.44	1.60	35	41.51	0.0001
O2M1C	4.53 ± 0.42				
A2M2H	3.30 ± 0.50	0.20	6	0.87	0.3727
O2M2H	3.50 ± 0.16				
A2M2C	1.90 ± 0.52	1.78	48	56.24	0.0001
O2M2C	3.68 ± 0.26				

*Note: Duration of light exposure is four times the manufacturer's recommendation for the Argon laser, i.e., 20 seconds. For the Optilux 401 VLC unit, it is one time the manufacturer's recommendation, i.e., 40 seconds.

TABLE 9. Temperature Rise (°C) at Pulpal Dentin Interface, Argon Laser vs. Optilux VLC Unit* (Continuous) (Argon Laser –20 sec vs. Optilux –80 sec)

Sample	Temp rise (20 vs. 80)	Difference	% Difference	*F* value	Pr > *F*
A1M	9.05 ± 0.91	4.03	31	86.19	0.0001
O1M	13.08 ± 0.55				
A1M1H	5.63 ± 0.19	0.70	11	28.35	0.0003
O1M1H	6.33 ± 0.26				
A1M1C	4.37 ± 0.49	1.98	31	94.05	0.0001
O1M1C	6.35 ± 0.10				
A1M2C	3.12 ± 0.27	3.38	52	145.91	0.0001
O1M2C	6.50 ± 0.63				
A2M	5.03 ± 0.30	1.79	26	13.39	0.0044
O2M	6.82 ± 1.16				
A2M1H	4.10 ± 0.33	2.93	41	417.76	0.0001
O2M1H	7.03 ± 0.12				
A2M1C	2.93 ± 0.44	3.65	55	144.39	0.0001
O2M1C	6.58 ± 0.60				
A2M2H	3.30 ± 0.50	2.23	40	97.00	0.0001
O2M2H	5.53 ± 0.24				
A2M2C	1.90 ± 0.52	3.93	67	261.41	0.0001
O2M2C	5.83 ± 0.29				

*Note: Duration of light exposure is four times the manufacturer's recommendation for the Argon laser, i.e., 20 seconds. For the Optilux 401 VLC unit, it is two times the manufacturer's recommendation, i.e., 80 seconds.

TABLE 10. Temperature Rise (°C) at Pulpal Dentin Interface, Argon Laser and Optilux VLC Unit, Continuous or Interrupted* (1-mm Dentin vs. 2-mm Dentin)

Sample	Max temp rise	Difference	% Difference	*F* value	Pr > *F*
A1M	9.05 ± 0.91	4.02	44	105.64	0.0001
A2M	5.03 ± 0.30				
PA1M	4.88 ± 0.85	1.73	35	21.25	0.0010
PA2M	3.15 ± 0.35				
O1M	15.00 ± 0.63	6.92	46	146.64	0.0001
O2M	8.08 ± 1.25				
PO1M	13.08 ± 0.70	6.85	52	326.57	0.0001
PO2M	6.23 ± 0.61				

*Note: Duration of light exposure is four times the manufacturer's recommendation. Argon laser: 5 seconds × 4; total light exposure was 20 seconds. Optilux 401 VLC unit: 40 seconds × 4; total light exposure was 160 seconds.

DISCUSSION

The argon laser has emerged as an alternative to conventional visible curing systems. One of the phenomena that accompanies the use of the argon laser in the photoinitiation and subsequent polymerization of resins is the generation of heat. This process and its relationship to the thermal tolerance of the pulp must be considered for clinical applications [2]. Because the argon laser emits fewer wavelengths of light over a narrow bandwidth specific to the absorption spectrum of the composite photoinitiator, argon laser curing of resin theoretically should offer greater time and energy efficiency. This greater efficiency should reduce the potential for extraneous energy absorption and heat build-up within the pulp compared with a conventional light [11]. Possible adverse effects of faster resin curing by using the laser, such as increase in polymerization shrinkage have not been adequately explored. Recently, Bouschlicher et al. in 1997 found reduced maximal polymerization shrinkage forces for laser versus conventional visible light curing of resin composite. However, these results were obtained in a nonrigid testing apparatus [8]. Losche and Roulet [12] noted that temperature rise within a tooth during argon lasing of resin was both time and power dependent. They also found that the thick-

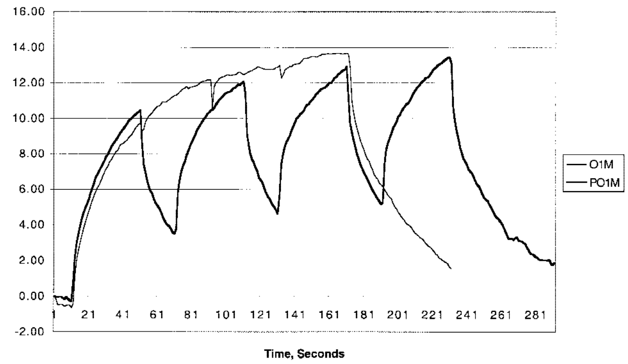

Fig. 2. Mean temperature change (°C) at the pulpal-dentin interface for both continuous and interrupted visible light curing exposure. Configuration used: 1-mm disk with neither a hole nor composite placed (O1M and PO1M).

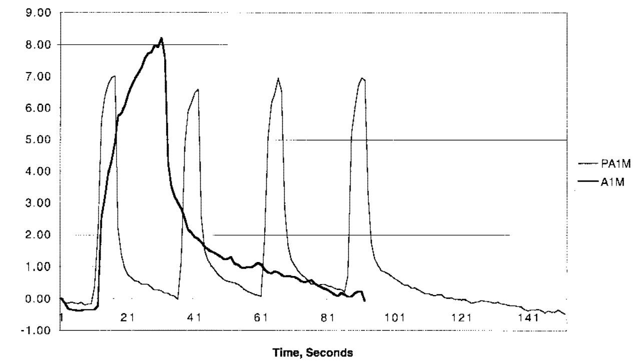

Fig. 3. Mean temperature change at the pulpal-dentin interface for both continuous and interrupted argon laser exposures. Configuration used: 1-mm disk with neither a hole nor composite placed (A1M and PA1M).

ness of the dentin barrier reduced the thermal increase as did the filtering of the argon ion laser beam to wavelengths less than 500 nm. The steepest temperature increase was recorded during the first 20 seconds. They concluded, with regard to temperature generation of an argon ion laser, that curing power should not exceed 500 mW.

The argon laser (wavelength, 457.9–514.5 nm) has been approved by the FDA for use in the oral cavity for soft-tissue procedures and curing of resin restorations. Studies have indicated its efficacy in caries prevention [13,14] and improving dentin adhesion [15,16]. Clinical applications use energy densities in the range of 25–100 J/cm^2. Heat generated during these procedures could inadvertently damage teeth or soft tissues if proper precautions are not taken or parameters for damage not understood [17].

Pulpal damage has been correlated previously with external surface incident energy density in the canine model [17]. This work suggests that the threshold energy density to be avoided is from 25–1,000 J/cm^2 and beyond. For comparison, curing a composite with a 250–500 mW argon laser for 10 seconds with an 8-mm diameter spot results in an energy density of 5–10 J/cm^2, or 12–25 J/cm^2 with a 5-mm spot. The authors then performed an in vitro correlation between histologic pulpal damage seen in the dog model with temperature data from the pulpal-dentin interface taken in vitro on both human and canine teeth. The authors found that teeth exposed to <600 J/cm^2 had pulpal histology similar to that of the control teeth. Slightly more than 600 J/cm^2 resulted in minimal damage to pulp tissue, which was assumed to be reversible. Over 800 J/cm^2 resulted in generalized necrosis of the pulp. The pulp wall temperature changes were minimal below 200 J/cm^2 of applied energy. Temperature increase was less than 6°C in human teeth when 900 J/cm^2 of laser energy was applied. These results were consistent with those of Cohen, who reported that pulpal damage begins with a temperature increase of 10°F (5.5°C) [18]. The argon laser parameters required to create a 10°F (5.5°C) pulpal temperature change in dogs' teeth were similar to the energy density that first caused pulpal damage in the histologic study of dogs' pulpal tissue changes. It is expected that, in humans, relatively greater energy densities would be necessary to cause pulpal damage, because of a greater thickness of enamel and dentin. From these results, the authors concluded that no apparent damage to the pulp or enamel of human teeth would be expected by using 100 mW to 1 W of power output for a laser energy density < 200 J/cm^2. Most applications of argon lasers for preventive and restorative dentistry use energy densities in the range of 25–100 J/ cm^2, well within this range.

Theoretically, the argon laser should be more efficient than a conventional light source for the photoinitiation of resin polymerization, because a greater proportion of its wavelengths are selectively absorbed by camphoroquinone. The conventional visible light system emits a much broader range of wavelengths, a small portion of which are effective for resin polymerization. Although filters reduce the emission of ineffective wavelengths, a portion of residual light energy produced from the broader light spectrum may still be transmitted to the target, potentially being absorbed to produce heat. In addition, the monochromaticity and collimation of the argon la-

ser light produces a focused beam, having a higher peak intensity than the divergent, attenuated light from conventional sources [19]. It may be that the greater portion of energy in effective frequencies increases both the degree and rate at which the diketone initiator camphoroquinone combines with the reducing agent to produce the free radicals necessary to initiate resin polymerization [1]. Early research supports this rationale. Results of early studies done on the physical properties and degree of polymerization of resins that used the argon laser compare favorably with those obtained from photoinitiation with a conventional visible light system [4]. Parameters for exposure times and power outputs were explored in several studies [17,19,20]. Some studies suggested that minimal exposure times were material (composite) specific. It was noted that both speed of polymerization and depth of cure could be effected by the characteristics of the material and the power of the laser. Variations in the actual type, concentration, and numbers of initiators between different brands of composite materials may explain this material specificity. Theoretically, a composite with multiple initiators may actually polymerize better with a multi-wavelength visible light curing unit than with a "wavelength-restricted" laser, if some of the initiators have activation wavelengths outside the primary laser bandwidth.

In 1965, Zach and Cohen [18] suggested that irreversible pulpal damage occurred in 15% of monkeys' teeth when pulpal temperature increased more than 5°C. More recent studies have supported this [12,17]. With a temperature rise over 11.1°C, virtually all the monkeys' teeth in vivo were found to undergo pulpal necrosis. However, Zach and Cohen used a soldering iron to generate the heat at the tooth surface, and this may not be directly comparable in a specific sense to the laser-generated heat in this study. Zach and Cohen essentially tested heat transfer by way of conduction from the surface whereas the present work tested heat transfer from both conduction and radiation throughout a volume of the tooth. It is also impossible to compare the energies delivered by Zach and Cohen and the laser in this study. However, it is possible that because of the capacitive nature (i.e., heat capacity or specific heat) of dentin, the duration of the temperature rise in the study by Zach and Cohen was sustained longer than in the present work, where a portion of the temperature increase was caused by radiation. Damage to tissue is a function of both temperature and duration (i.e., energy). This finding suggests that the temperature thresholds described by Zach and Cohen may in fact be lower than the actual laser-induced temperature thresholds.

The worst-case VLC unit and argon laser temperature increases occurred with a 1-mm disk and maximal continuous exposure. The worst-case VLC temperature of 15.00°C exceeded the threshold temperature described by Zach and Cohen (11.1°C). The worst-case argon laser temperature of 9.05°C approached this temperature. Notwithstanding the previous comments, histologic evidence is still needed to determine the pulpal effect of this or any recommended laser exposure to teeth. Under the same exposure conditions, through 2 mm of dentin, temperature rise was reduced by almost half for both light sources, 5.03°C for the argon and 8.08°C for the VLC unit. Thus, dentin seems to be an effective insulator. In both cases, the risk to the pulp seems less with the argon laser than with the VLC unit. The insulating effect of increasing thickness of remaining dentin between the light source and the pulp corroborates several earlier studies [12,21,22].

Other investigators have looked at the contribution of exotherm from composite polymerization to the overall pulpal temperature rise and have even quantified this factor. Within the limits of this study, polymerization exotherm did not prove to be a significant factor in contributing to the total temperature increase in most cases. In specimens for which composite was polymerized during exposure to the VLC unit, no consistently significant increase in pulpal temperature was found compared with the same tooth configuration without composite. This finding may have been due to the fact that the total heat from the light source was overwhelming compared with the exotherm from the small restoration that was polymerized. When the total temperature rise caused by the light was small, a different pattern emerged. Exposure to the argon laser through at least 3 mm dentin by using the interrupted cycle produced the highest temperature elevation during the third exposure cycle (when composite was cured) in three of the four groups. For other groups, the highest temperature rise occurred at the conclusion of the total light exposure. This suggests that the composite exotherm made a contribution to overall temperature elevation in those cases for which composite was cured by using the argon laser during interrupted cycles.

CONCLUSIONS

Within the limits of this study, the argon laser consistently caused a smaller maximal temperature rise than did the VLC unit. Other factors also affected temperature increase. Dentin thickness was inversely proportional to temperature rise. Continuous exposure to either light source produced higher temperature increases than did similar exposure times by using the interrupted cycle. Exotherm from resin polymerization did not seem to make a significant contribution to temperature rise. The temperature rise was initially rapid and diminished as thermal equilibrium is approached, tapering off at the end of the heating cycle.

REFERENCES

1. Blankenau RJ, Powell GL, Kelsey WP, Barkmeier WW. Post polymerization strength values of an argon laser cured resin. Lasers Surg Med 1991;11:471–474.
2. Sakaguchi R, Douglas WH, Peters MC. Curing light performance and polymerization of composite restorative materials. J Dent Res 1992;20:183–188.
3. Hansen EK, Asmussen E. Correlation between depth of cure and surface hardness of a light-activated resin. Scand J Dent Res 1993;101:99–102.
4. Blankenau RJ, Kelsey WP, Powell GL, Shearer GO, Barkmeier WW, Cavel WT. Degree of composite resin polymerization with visible light and argon laser. Am J Dent 1991;4:40–42.
5. Kelsey WP, Blankenau RJ, Powell GL, Barkmeier WW, Whisenant BK. Enhancement of physical properties of resin restorative materials by laser polymerization. Lasers Surg Med 1989;9:623–627.
6. Cobb DS, Vargas MA, Rundle T. Physical properties of composites cured with conventional light or argon laser. Am J Dent 1996;9:199–202.
7. Vargas MA, Cobb DS, Schmit JL. Polymerization of composite resins: argon laser vs. conventional light. Oper Dent 1998;23:87–93.
8. Bouschlicher MR, Vargas MA, Boyer DB. Effect of composite type, light intensity, configuration factor and laser polymerization on polymerization contraction forces. Am J Dent 1997;10:88–96.
9. Trowbridge H, Franks M, Korostoff E, Emling R. Sensory response to thermal stimulation in human teeth. J Endod 1980;6:405–412.
10. Waknine S, Cipolla AJ. Laser vs. halogen photopolymerization-characterization of VLC composite mechanical properties. J Dent Res 1995;74:230.
11. Goodis H, White J, Gaum B, Watanabe L. Pulp chamber temperature changes with visible light-cured composites in vitro. Dent Mater 1990;6:99–102.
12. Losche G, Roulet J. Temperature rise by visible light curing units and argon laser. J Dent Res 1991;70:362.
13. Hicks MG, Flaitz CM, Westerman GH, Blankenau RJ, Powell GL, Berg JH. Enamel caries initiation and progression following low fluence (energy) argon laser and fluoride treatment. J Clin Pediatr Dent 1995;20:9–13.
14. Hicks MJ, Flaitz CM, Westerman GH, Blankenau RJ, Powell GL. Root caries in vitro after low fluence argon laser and fluoride treatment. Comepend Contin Educ Dent 1997;18:543–548.
15. Shanthala BM, Munshi AK. Laser vs visible-light cured composite resin: an in vitro shear bond study. J Clin Pediatr Dent 1995;19:121–125.
16. Powell GL, Blankenau RJ. Effects of argon laser curing on dentin shear bond strengths. J Clin Laser Med Surg 1996;14:111–113.
17. Powell G, Morton T, Whisenant B. Argon laser oral safety parameters for teeth. Lasers Surg Med 1993;13:548–552.
18. Zach L, Cohen G. Pulp response to externally applied heat. Oral Surg 1965;19:515–530.
19. Kelsey WP, Blankenau RJ, Powell GL, Barkmeier WW, Stormberg EF. Power and time requirements for use of the argon laser to polymerize composite resins. J Clin Laser Med Surg 1992;10:273–278.
20. Burtscher P. Curing of composites with an argon laser. J Dent Res 1991;70:526.
21. White JM, Fagan MC, Goodis HE. Intrapulpal temperatures during pulsed Nd:YAG laser treatment of dentin, in vitro. J Periodontol 1994;65:255–259.
22. Stewart G, Bachman B, Hatton J. Temperature rise due to finishing of direct restorative materials. Am J Dent 1991;4:23–28.